The Effect of Veneering Materials on Stress Distribution in Implant-Supported Fixed Prosthetic Restorations

Yalçın Çiftçi, DDS, MS[1]/Şenay Canay, DDS, MS[2]

In this study, the effect of various materials used in fabricating superstructures for implant-retained fixed partial dentures on stress distribution around implant tissues was investigated. Five different mathematical models consisting of 11,361 nodes and 54,598 elements were constructed to study porcelain, gold alloy, composite resin, reinforced composite resin, and acrylic resin veneering materials using the 3-dimensional finite element analysis method. MARC K7.2/Mentat 3.2 software was used for the analysis. Reference points were determined on the cortical bone, where perpendicular, oblique, and horizontal forces were applied. Stress values created by oblique and horizontal forces appeared to be higher than those created by vertical forces. Stress seemed to be concentrated at the cortical bone around the cervical region of the implant. Gold alloy and porcelain produced the highest stress values in this region. Stresses created by acrylic resin and reinforced composite resin were 25% and 15% less, respectively, than porcelain or gold alloy. Porcelain and gold alloy produced stress values at the lingual implant sites that reached the ultimate strength values of the cortical bone. (INT J ORAL MAXILLOFAC IMPLANTS 2000;15:571–582)

Key words: *dental implant, dental materials, dental stress analysis, finite element analysis, fixed partial denture*

Biomechanical factors play an important role in the long-term survival of oral implants. The selection of implant positions, prosthesis design, and superstructure material is critical for the longevity and stability of the implant prosthesis.[1–7] The nature and magnitude of loads necessary to cause implant loosening are unknown, so it has been recommended that forces be kept to a minimum. The osseointegrated implant provides direct contact with bone and has no micromovement. Therefore, all stress waves or shocks applied are transmitted to the implants.

Because of the lack of micromovement of osseointegrated implants, most of the force distribution is concentrated on the crest of the ridge, and this may lead to bone resorption and subsequent loss of the implant. It has been suggested that stress-absorbing or load-dampening systems be incorporated into the superstructures supported by osseointegrated implants, so as to reduce loads on the implant that occur because of the lack of viscoelasticity at the bone-implant interface.

The IMZ System (Interpore International, Irvine, CA) has an internal shock absorber made of a medical-grade polyoxymethylene incorporated into its design.[8–10] The manufacturer of the Brånemark System (Nobel Biocare, Göteborg, Sweden) has recommended acrylic resin for fabrication of the occlusal surfaces of prostheses.[11] Other researchers have demonstrated that acrylic resin provides a generous layer of material with sufficient cushioning effect to dampen most commonly exerted oral forces and thus acts as a shock absorber.[12–14] However, acrylic resin wears at an accelerated rate when opposed by natural teeth or porcelain.[15,16] Such changes in the occlusal surfaces will reduce chewing efficiency and may alter maxillomandibular relationships.[17] This has led authors to encourage the use of porcelain on occluding surfaces.[18]

[1]Assistant Professor, Department of Prosthodontics, Faculty of Dentistry, University of Hacettepe, Ankara, Turkey.
[2]Associate Professor, Department of Prosthodontics, Faculty of Dentistry, University of Hacettepe, Ankara, Turkey.

Reprint requests: Dr Şenay Canay, Bağiş sok. 17/10, 06660 Kocatepe Ankara, Turkey. Fax: 90 (312) 4184715. E-mail: yalcin@optima.com.tr

Figs 1a to 1d Mathematical model including implants and superstructures.

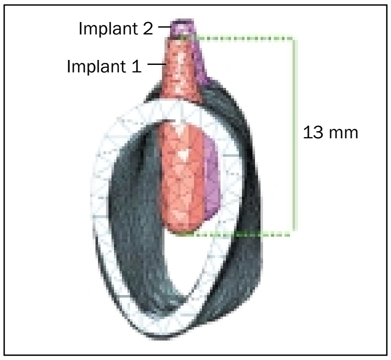

Fig 1a Section with implants and cortical bone.

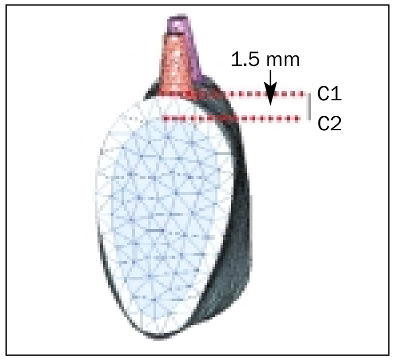

Fig 1b Section with implants, cortical bone, and trabecular bone. C1 = crest of cortical bone; C2 = junction between cortical and cancellous bone.

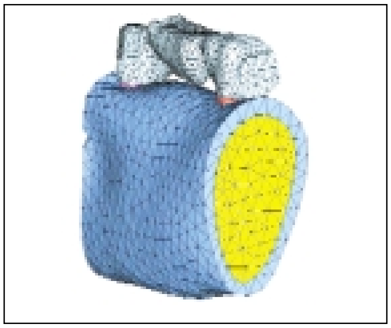

Fig 1c Section with implants, cortical bone, trabecular bone, and copings.

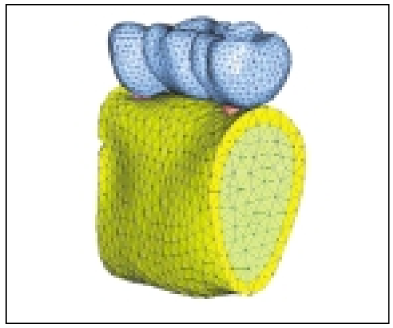

Fig 1d Section with implants, cortical bone, trabecular bone, copings, and veneering materials.

Ceramic materials are acceptable for veneering implant-supported prostheses, and their use greatly improves the esthetics of implant restorations.[19] However, porcelain is not a stress-absorbing material, so that forces developed at the occlusal level will be transmitted directly to the prosthesis, the implant, or the bone interface unless they are mediated in some fashion.[10,20] The intensity of the resulting stresses would be a function of the physical qualities of the different veneering materials used. The thickness of the veneering material is greatly reduced when a partially edentulous arch is restored. The question arises as to whether or not resin used as a veneering material provides a significant cushioning effect.

The finite element analysis (FEA) method has proved to be a useful tool in estimating stress levels around implants. It involves the development of a mathematical model of a continuous structure divided into a system of discrete components or elements. These components are connected at nodal points, where stresses and displacements are determined. The accuracy of the 3-dimensional (3-D) method is proportional to the number of nodes and elements in the mathematical model.[21,22]

The purpose of this study was to analyze, with the 3-D FEA method, the force-absorbing behavior of 5 restorative materials used to veneer superstructures that were rigidly connected to implants.

MATERIALS AND METHODS

In this study, the FEA method was used to evaluate stresses in the mandibular posterior quadrant, where an implant-supported fixed partial prosthesis was fabricated with different restorative materials. The implants were assumed to be placed in the region of the second premolar and second molar, and the first molar tooth was assumed to be lost. The model was provided with 2 spline hydroxyapatite-coated cylindric Calcitek dental implants (Sulzer Calcitek Inc, Carlsbad, CA) 4 mm in diameter and 13 mm in height. A 3-D mathematical model including a framework, abutments, implants, a bone-implant interface, and a section of the mandible was constructed, representing the anatomic geometry of the mandible (Fig 1).

An actual human mandible was used in the preparation of mathematical models. With the help of laser scanners (3D Digitizer Model 3030 and Echo digitizing software, resolution 1 × 1 mm/30 s, Cyberware, Monterey, CA), the surface topography of prosthetic superstructures and mandibles was converted into digital data. Data were converted into a 3-D solid mathematical model by using the solid and crest modeling programs I-DEAS Artisan Series 3 (Structural Dynamics Research Corporation, Milford, OH) and the MARC K7.2/Mentat 3.2 FEA program (MARC Analysis Research Corporation, Palo Alto, CA). For the analysis, a direct profile was used. In this procedure, the cores of shell structures were meshed with 4-node tetrahedron elements. The analysis program used in the study could easily automesh the complicated model when the tetrahedron elements were used. Since these nodal points could move in 3 axes, their degree of freedom was set at 3. The displacement of these nodes assisted in the calculation of stress distribution inside the structures.

To simulate the clinical situation, the model was not supported at the bottom; instead, it was fixed along the points where the masticatory muscles were inserted. A pilot study (ie, a trial run) showed that the region where the analyses were done was not influenced by the location of the muscle insertions. Therefore, the mesh was refined in the area of implant placement to provide additional stress resolution in this region. The mathematical model was divided into 54,598 elements connected at 11,361 points, known as nodes.

Mandibular anatomy mandated placement of the posterior implant with a lingual inclination to take advantage of the bone height in the model, with moderate (10-degree) inclination. The modeled section of the mandible had an 18-mm buccolingual thickness, was 26 mm in height, and was surrounded by 1.5 mm of cortical bone.

A superstructure that represented the framework of an implant-supported fixed prosthesis was also modeled at 4 mm in height and 6 mm in width. The materials used as veneering materials were Type III gold alloy (Sjoding C-3); feldspathic porcelain (Vita VMK 68, Vident, Brea, CA); heat-polymerized polymethylmethacrylate resin (Biotone, Dentsply Co, York, PA); microfilled composite resin (Charisma, Heraeus Kulzer, South Bend, IN); and glass-modified composite resin (Artglass, Heraeus Kulzer). Restorative materials varied, but the abutment designs were similar, so that the resulting stress and stress distribution could be attributed to the material differences. The cement layer between the crown and abutment was too thin to adequately model in the finite element simulation and was considered to be negligible.

Table 1 Mechanical Properties of Oral Tissues and Prosthetic Materials in FEA Evaluations

	Modulus of elasticity GPa	Poisson's ratio (v)
Cortical bone	13.7	0.30
Cancellous bone	1.37	0.30
Titanium	117	0.33
Metal coping (Ceramco O)*	86.2	0.33
Full-crown Type III gold alloy (Sjoding C-3)	100	0.33
Porcelain (Vita VMK 68)	70	0.19
Glass-modified composite resin (Artglass)	10	0.30
Acrylic resin (Biotone)	2.26	0.35
Composite resin (Charisma)	14.1	0.24

*Ceramco/Dentsply, Weybridge, Surrey, England.

All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. To simulate ideal osseointegration, the implants were rigidly anchored along their entire interface in the bone model. Modulus of elasticity and Poisson's ratio values are presented in Table 1.

A wide range of magnitudes for chewing forces has been reported in the literature. For the current model, 3 forces from different directions were selected: a horizontal bite force (F_h = 0 degrees), a vertical bite force (F_v = 90 degrees), and an oblique bite force (F_o = 120 degrees). The proportion of the force magnitude was $F_h : F_v : F_o = 1 : 3.5 : 7$.[23] In the study, a vertical load of 500 N, a horizontal load of 142 N, and an oblique load of 1,000 N were applied.[24] The vertical and oblique loads were applied equally at the 125 nodal points on the buccal inclination of the lingual cusps. Hence, when the mandibular implant opposes a natural maxillary tooth, the primary contacting cusp becomes the maxillary lingual cusp opposing the mandibular implant crown, with the mandibular buccal cusp of decreased height and width over the implant body.[25] The horizontal loads were applied to the nodal points on the buccal.

Loading forces on the models were static. Stress contours were computed and plotted in the bone tissue. Then by giving a 0.1-mm displacement to the nodes defined in a specific period of time, stress distributions formed on the cortical bone were analyzed.

Before loading, specific points at 2 levels along the bone-implant interface were selected for convenience to directly compare models representing different analysis variables.

1. The crest of the cortical bone (C1). Three points were chosen at the buccal and lingual aspects of both the first and second implants, which were placed in the second premolar and second molar region, as reference (Fig 1b).
2. The junction between the cortical and cancellous bone (C2). Two points were chosen for the second implant, and 3 points were chosen for the first implant on the buccal and lingual aspects (Fig 1b).

For an evaluation of stress distribution, the magnitudes of the concentrations were presented in principal stresses. The principal stress offers the possibility of making a distinction between tensile stress and compressive stress. Displacement components of specific points provide information about the deformation of the model and facilitate interpretation of the results. The magnitude of the stresses from these reference points was evaluated for each of the 5 different veneering materials. Principal stress values of fragile compact bone were compared with the ultimate compressive strength and ultimate tensile strength values.

The use of statistical analyses was very limited for the FEA studies because the results of the model observations were invariant.

RESULTS

Stress patterns appeared as contour lines with different color connecting equistress points between certain limits. The stress value for each contour line was presented as a positive or negative; positive values identified tensile stresses, while negative values identified compressive stresses. For an evaluation of stress distribution, the magnitude of the concentrations was presented in minimum (compressive stress) and maximum (tensile stress) principal stress, together with their location in relation to the implant. Maximum and minimum principal stress values that resulted from vertical and oblique loading conditions on the bone-implant interfaces and the cortical bone surrounding the neck of the implants with 5 different veneering materials are illustrated in Figs 2 to 5. Stress contours were color-coded and were explained on the left side of each figure for both loading magnitudes. Table 2 summarizes the magnitude and type of stresses.

The intensity of compressive stress was higher when vertical and horizontal loads were transferred to the upper and lower crests of the cortical bone found at the first implant's lingual surface. These values were highest for gold alloy and lowest for the acrylic resin veneering material. Lower values were also found for composite resin and glass-modified composite resin. All the stress values seen at the C2 level of the cortical bone show that there was a reduction in stress from the C1 level. However, when oblique and horizontal loads were transferred to the C1 and C2 level of the cortical bone on the buccal crest, compressive stresses were more likely to be formed.

In this study, tensile and compressive stresses were evaluated. However, in the regions where the maximum principal and minimum principal stress values were similar, the stresses could not be defined as actually tensile or compressive; therefore the term *complex* or *mixed* stress was used. On the C1 level of the cortical bone, when vertical load was transferred, complex stresses were formed only on the gold alloy and porcelain. In contrast, when acrylic resin, composite resin, and glass-modified composite resin materials were used, tensile stress was more likely to occur. Tensile and compressive stresses that existed on the second implant's lingual and buccal aspects were greater than those on the first implant's lingual and buccal sites. This was true for all 5 veneering materials under all 3 loading conditions. Higher stresses were situated in the upper cortical layer than in the lower regions. Gold alloy had the highest average tensile stress values observed under vertical, oblique, and horizontal loads, but with porcelain, composite resin, glass-modified composite resin, and acrylic resin, this rate decreased gradually. On the buccal aspect of the cortical bone, stresses were complex in nature under vertical loading but were predominantly tensile under horizontal and oblique loading.

Cortical Bone Displacement Results

In the study, static analysis findings were considered, and stress rates under applied loads were recorded. After the same amount of load was applied vertically on the prespecified nodal points, the highest stress levels that formed on the upper crest of the cortical bone around both implants were observed in the models where gold alloy and porcelain were used as veneering material, and the lowest stress levels were observed in the model in which acrylic resin was used. Similar conclusions were drawn for the lower crest of the cortical bone, which showed a decrease in the stress level compared to that of the upper crest (Figs 6 and 7).

Figs 2a to 2e Maximum principal stress values (MPa) around implants restored with 5 different veneering materials under vertical loading.

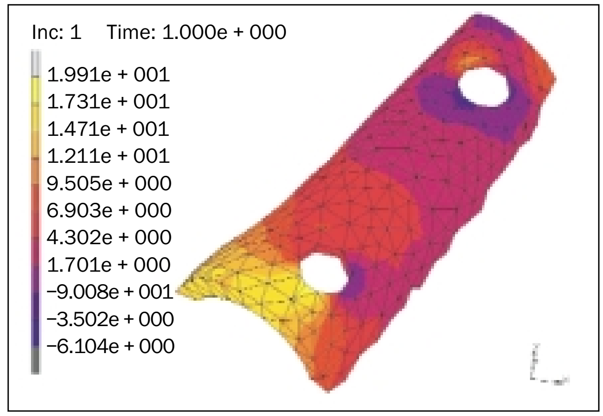

Fig 2a Porcelain.

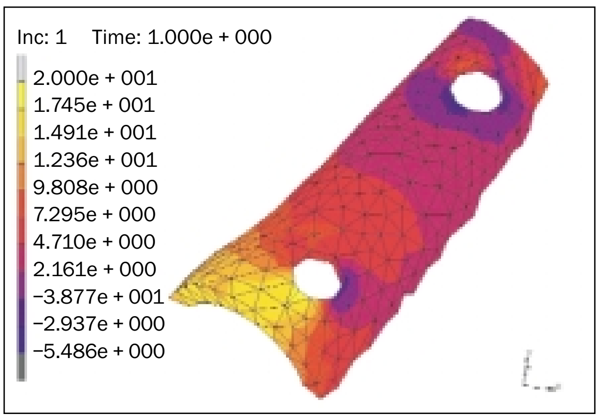

Fig 2b Glass-modified composite resin.

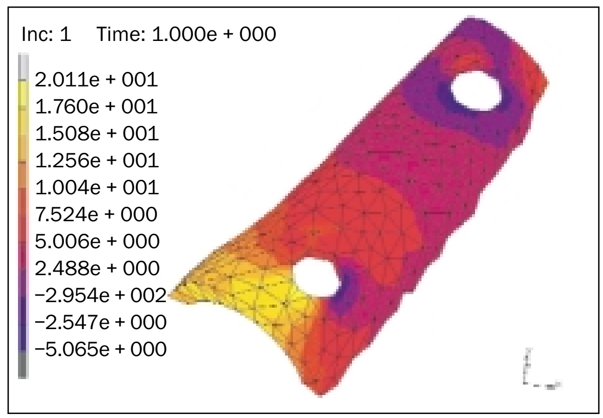

Fig 2c Acrylic resin.

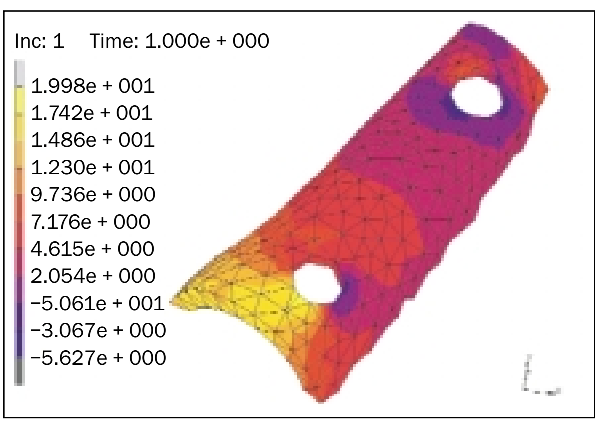

Fig 2d Composite resin.

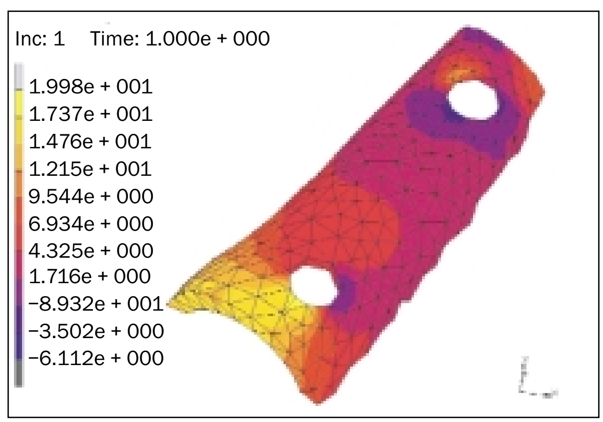

Fig 2e Gold alloy.

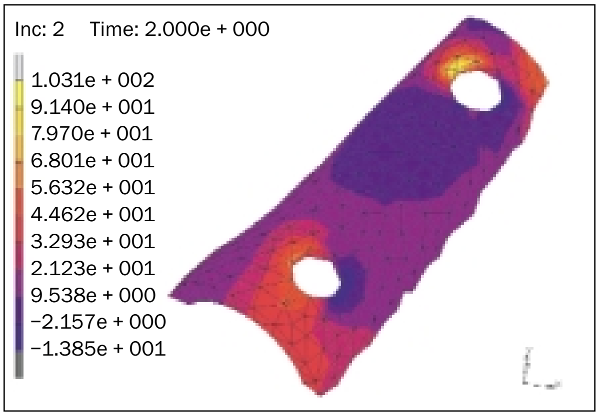

Fig 3a Porcelain.

Figs 3a to 3e Maximum principal stress values (MPa) around implants restored with 5 different veneering materials under oblique loading.

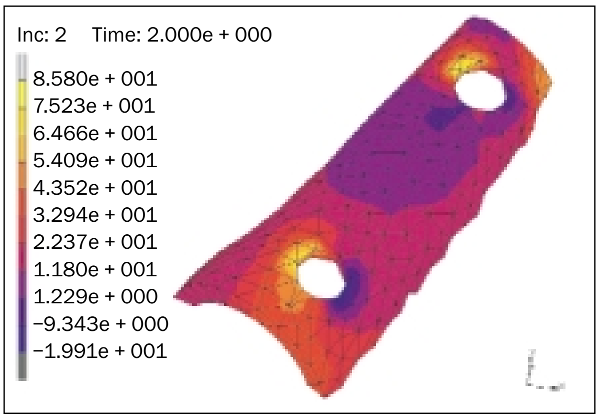

Fig 3b Glass-modified composite resin.

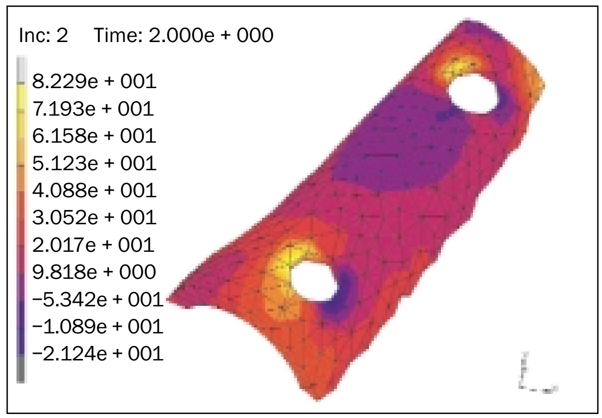

Fig 3c Acrylic resin.

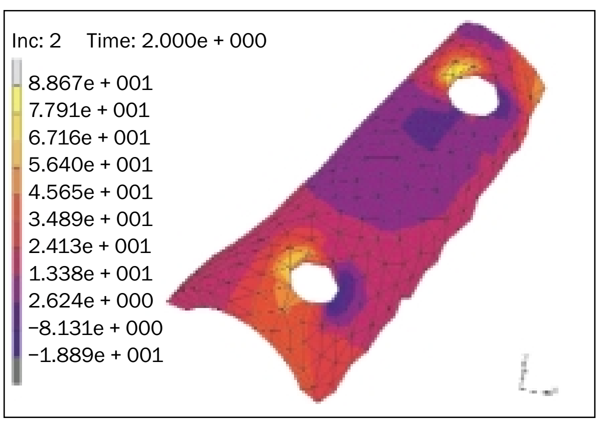

Fig 3d Composite resin.

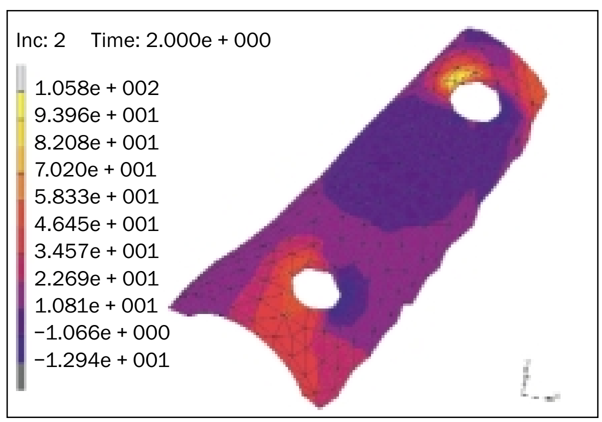

Fig 3e Gold alloy.

Figs 4a to 4e Minimum principal stress values (MPa) around implants restored with 5 different veneering materials under vertical loading.

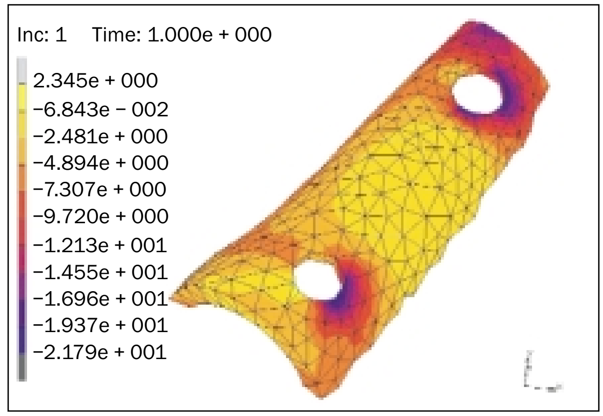

Fig 4a Porcelain.

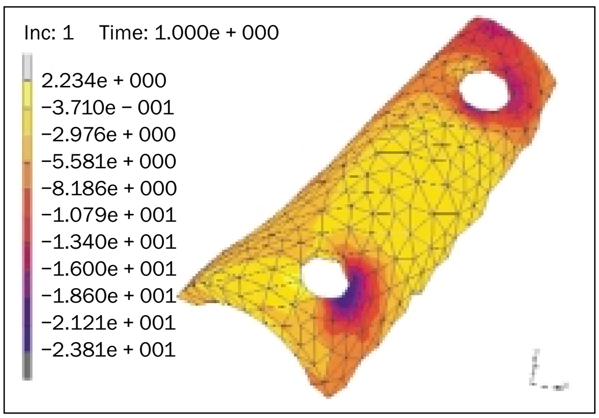

Fig 4b Glass-modified composite resin.

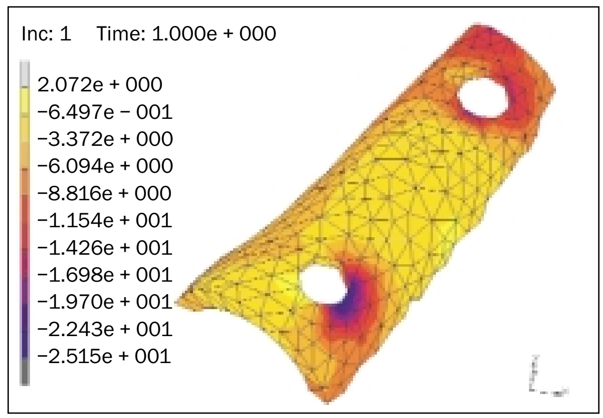

Fig 4c Acrylic resin.

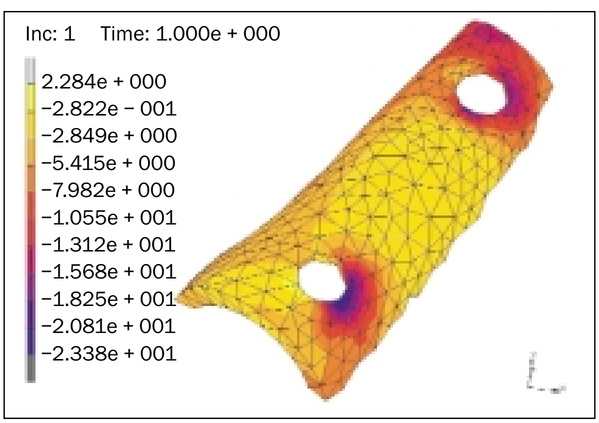

Fig 4d Composite resin.

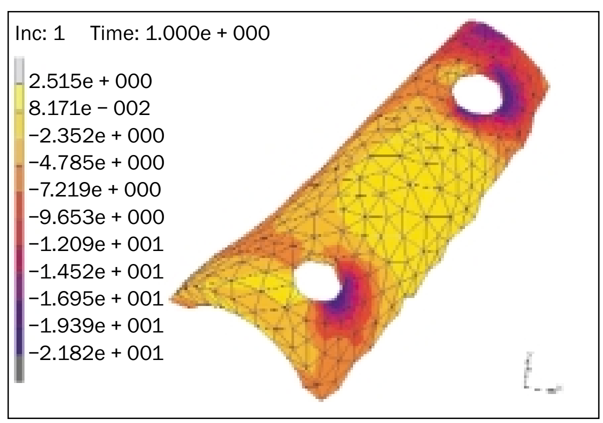

Fig 4e Gold alloy.

Figs 5a to 5e Minimum principal stress values (MPa) around implants restored with 5 different veneering materials under oblique loading.

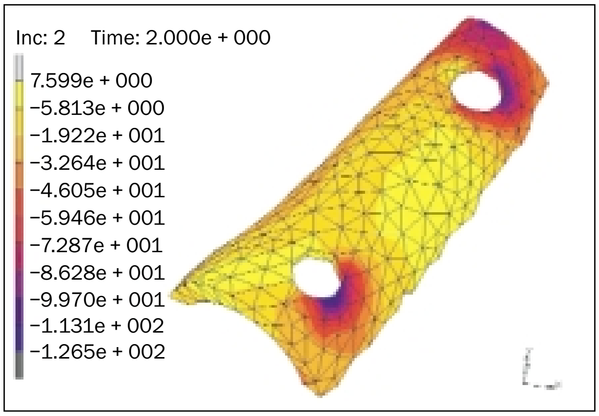

Fig 5a Porcelain.

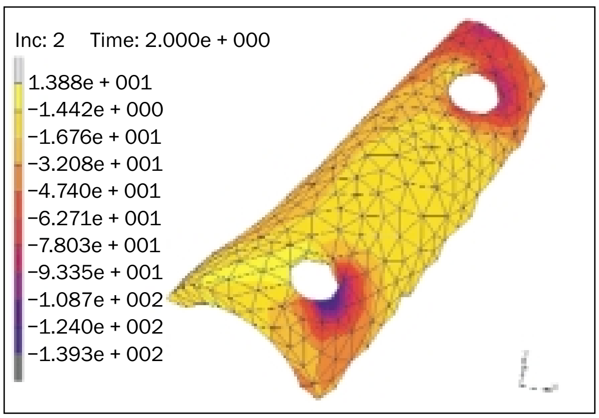

Fig 5b Glass-modified composite resin.

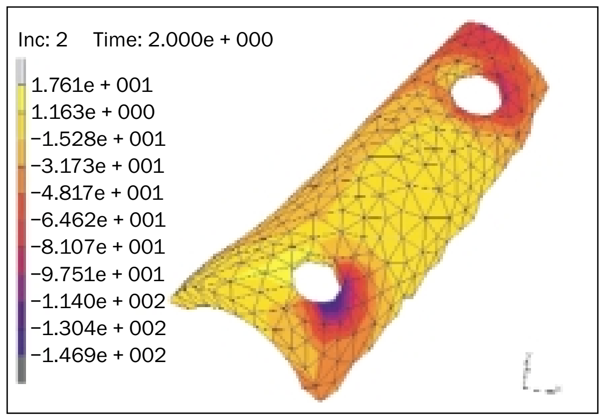

Fig 5c Acrylic resin.

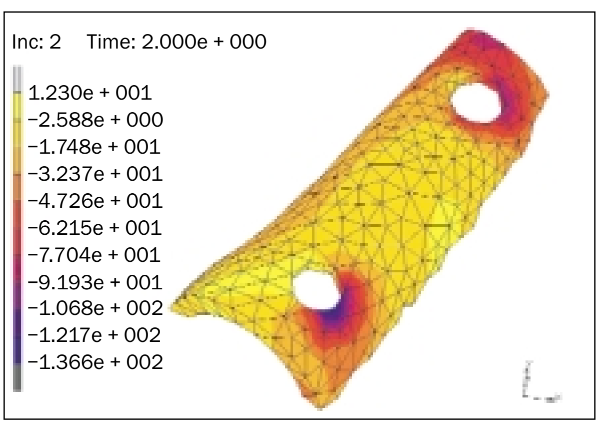

Fig 5d Composite resin.

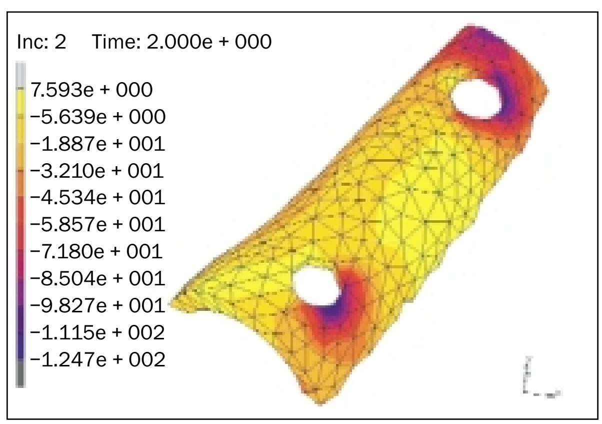

Fig 5e Gold alloy.

Table 2 Magnitude and Type of Principal Stresses around the First and Second Implants at C1 and C2 Levels

	Maximum principal stress (MPa)						Minimum principal stress (MPa)						Type of stress					
	Vertical		Oblique		Horizontal		Vertical		Oblique		Horizontal		Vertical		Oblique		Horizontal	
	L	B	L	B	L	B	L	B	L	B	L	B	L	B	L	B	L	B
Implant 1																		
Level C1																		
Porcelain	–0.60	4.55	–0.20	54.10	0.50	64.15	–17.75	–6.15	–86.50	–22.75	–95.80	–24.85	C	M	C	T	C	T
Glass-modified composite resin	–0.55	2.75	2.40	45.95	3.70	55.35	–15.50	–7.25	–78.05	–24.50	–86.45	–28.25	C	C	C	T	C	T
Acrylic resin	–0.45	1.95	5.35	40.25	7.10	49.10	–10.55	–7.85	–64.80	–21.50	–74.55	–26.15	C	C	C	T	C	T
Composite resin	–0.50	3.00	1.80	47.55	2.85	57.00	–15.90	–7.10	–79.55	–26.85	–88.20	–29.30	C	C	C	T	C	T
Gold alloy	–0.55	4.95	–0.20	55.35	0.10	65.45	–18.00	–6.00	–87.60	–21.60	–97.00	–23.45	C	M	C	T	C	T
Level C2																		
Porcelain	1.33	4.70	9.47	50.00	11.00	58.50	–12.27	–4.00	–66.67	–17.10	–74.73	–18.70	C	M	C	T	C	T
Glass-modified composite resin	1.47	3.20	10.13	42.50	11.60	51.50	–11.00	–4.70	–61.53	–22.00	–69.13	–24.50	C	M	C	T	C	T
Acrylic resin	1.40	2.30	10.27	38.10	11.73	45.40	–10.33	–3.90	–55.60	–18.70	–66.00	–21.40	C	M	C	T	C	T
Composite resin	1.40	3.50	10.07	43.80	11.47	51.60	–11.20	–4.60	–62.40	–21.40	–70.00	–23.70	C	M	C	T	C	T
Gold alloy	1.27	5.00	10.27	51.10	11.93	59.60	–12.40	–3.90	–67.47	–16.10	–75.60	–17.60	C	M	C	T	C	T
Implant 2																		
Level C1																		
Porcelain	2.53	10.27	–2.13	63.40	–9.33	76.07	–20.13	–1.33	–120.80	6.93	–140.20	5.60	C	T	C	T	C	T
Glass-modified composite resin	2.80	9.20	1.40	55.00	1.00	66.33	–17.87	–2.93	–105.80	1.73	–123.13	0.47	C	T	C	T	C	T
Acrylic resin	2.00	10.93	–11.27	49.67	–13.47	64.33	–17.60	–0.40	–96.34	9.53	–117.00	8.87	C	T	C	T	C	T
Composite resin	2.80	9.47	0.80	56.33	0.33	67.67	–18.20	–2.67	–111.20	–2.33	–124.80	–5.07	C	T	C	T	C	T
Gold alloy	2.40	7.40	–8.73	64.60	–10.40	76.40	–20.60	–1.60	–127.13	8.80	–142.60	10.20	C	T	C	T	C	T
Level C2																		
Porcelain	3.00	6.50	8.40	35.90	8.80	43.40	–13.90	–3.70	–70.80	–1.40	–78.60	–1.50	C	M	C	T	C	T
Glass-modified composite resin	2.90	5.80	6.60	31.00	6.90	37.80	–12.40	–4.60	–64.20	–4.70	–71.20	–4.90	C	M	C	T	C	T
Acrylic resin	3.00	3.47	9.80	36.24	10.40	31.2	–11.60	–2.9	–56.40	–3.1	–61.30	–1.9	C	M	C	T	C	T
Composite resin	3.00	6.10	6.70	31.70	7.00	38.70	–12.70	–4.40	–65.20	–4.00	–72.30	–4.20	C	M	C	T	C	T
Gold alloy	3.00	6.60	8.70	36.80	9.30	44.40	–14.30	–3.50	–72.10	–1.10	–80.00	–1.60	C	M	C	T	C	T

L = lingual; B = buccal; C = compressive stress; T = tensile stress; M = mixed stress.

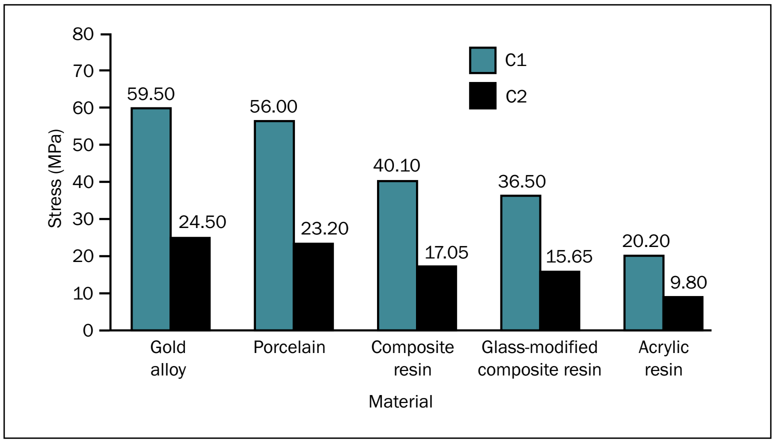

Fig 6 Normal principal stress values (for the displacement) around the first implant at both C1 and C2 levels.

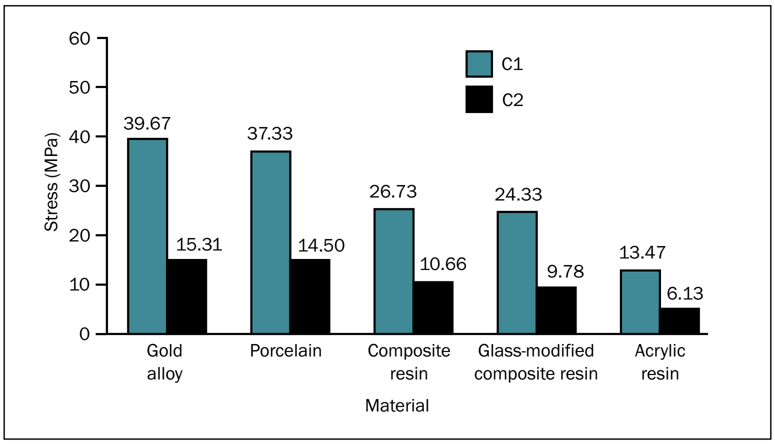

Fig 7 Normal principal stress values (for the displacement) around the second implant at both C1 and C2 levels.

DISCUSSION

This study used the 3-D FEA method to compare stress distribution in a mandibular posterior segment in which an implant-supported prosthesis was fabricated using different types of veneering materials. The bone quality and quantity in the cervical region may be critical to the long-term success of dental implants. Its loss could endanger implant stability. The type of veneering material is important for conducting the stress generated by static or impact forces to the lower structures.[5,12,25]

Since the modulus of elasticity of the composite resin (14.1 GPa) is higher than that of the glass-modified composite resin and acrylic resin, this material is resistant to fracture during mastication and absorbs maximum energy. On the other hand, porcelain transfers nearly all of the load to the bone because of its higher modulus of elasticity (70 GPa). Glass-modified composite and acrylic resins, with a lower modulus of elasticity (10 and 2.26 GPa, respectively), transfer minimal load to the bone and absorb the load.

When findings from this study were analyzed, it was seen that stress levels to the cortical bone around implants were lower in the models that used acrylic resin, composite resin, or glass-modified composite resin as veneering materials compared to those with porcelain or gold alloy. Davis and colleagues[12] concluded that acrylic resin helps to reduce the stresses under impact conditions, such as those that occur when a patient inadvertently bites on a hard object. Gracis and coworkers[14] used strain gauges in their study and concluded that microfilled light-cured and heat-cured composite resins and heat-cured polymethylmethacrylate resin materials showed impact forces that were 50% lower than those of porcelain and or gold alloy. Brånemark[11] recommended acrylic resin teeth for occluding surfaces of complete-arch prostheses in completely edentulous patients treated with implants. He inferred that this material would compensate for the resiliency normally provided by the periodontium. However, other researchers concluded that changing the veneering material on the prosthesis had no significant effect on the stress level or distribution at the bone-implant interface.[5,26]

According to the present results under static loading and impact conditions, the use of acrylic resin reduces the stress that is transmitted to the framework and the cortical bone. Another advantage of acrylic resin is the relative ease with which it can be added to the framework and adjusted when necessary. However, low resistance to abrasion and fracture are disadvantages to its use.

When the structured occlusal scheme and morphology cannot be maintained over time, undesirable lateral forces may increase.[17,27] Glass-modified composite resins represent an attempt to eliminate the disadvantages of acrylic and composite resins. Glass-modified composite resins provide wear resistance similar to tooth enamel. Inorganic active silica and microparticles of barium, aluminum, and silicon glass (Ba-Al-Si glass) were found among the glass particles, which were higher in organic content. Thus glass-modified composite resin is stiffer and more resistant to fracture.

In light of the high stress calculated for oblique and horizontal forces in each model, it is important to create an occlusal scheme that minimizes lateral force components in planning and fabricating a superstructure. Buccal cusp inclination was set at 30 degrees in the present models, and force was equally distributed at 125 predetermined nodal points on the buccal inclination of the lingual cusps. The ultimate tensile and compressive strength values of cortical bone have been reported as 121.00 MPa and 167.00 MPa, respectively.[28] In models that used metal as the veneering material, compressive stress values of a maximum of 119.40 MPa were measured on the lingual side of the cortical bone of the first implant and 157 MPa at the second implant. These values are very close to the ultimate strength value of the cortical bone, which is 167 MPa.

This study has demonstrated that stress values found at the C1 and C2 level of bone around the implant are a function of the veneering material. The compressive stress values on the lingual aspect of the cortical bone around the second implant were higher than those in the same region of the first implant. This may be because of the 10-degree lingual inclination of the second implant resulting from the natural angulation of the mandible.

Impact forces have more destructive effects on the bone surrounding the implants and on the superstructures. Stress distribution is directly related to the elastic modulus of the veneering material, ie, the stiffer the material, the more stress will be transmitted to the bone.

A properly selected prosthetic material will minimize forces on implants and consequently reduce stresses in the supporting bone. It is impossible to find every desirable characteristic in a single veneering material. Acrylic resin reduces impact force; however, it absorbs water easily, which contributes to discoloration. Porcelain has a higher modulus of elasticity than acrylic resin, so if it is added to a framework, as compared to acrylic resin or composite resin, more stress will be taken by the superstructure (under a static load). The greater wear resistance of porcelain, which may contribute to premature overload on the implants over time, would indicate a need for cautious use. Moreover, opposing dentition, as well as the personal characteristics of the patient and potential parafunctional mandibular movements, should be noted.

When metal alloy was used as the superstructure material, similar characteristics to porcelain in terms of load transfer to the implant and its surroundings were seen; however, it is esthetically inappropriate. In contrast, composite resins or glass-modified composite resins allow the formation of low stress levels in the bone around the implant, and their esthetic qualities are very satisfactory. Particularly with the Artglass system, its increased bonding with the metal framework by the Kevloc system (Hereaus Kulzer), ease of implant placement and working in the mouth, and the present findings of minimal stress levels under all types of loading provide significant advantages. Glass particles incorporated into the structure add resilience. Thus, it is an alternative material that can be used comfortably for implant-supported fixed partial prostheses.

CONCLUSIONS

The mechanical behavior of an implant-supported framework for which 5 different veneering materials were used was examined using FEA. The following conclusions were drawn:

1. For all veneering materials, stress was highest under horizontal and oblique loading and lowest under vertical loading.
2. For all models, the most extreme stress values were located within the implant collar immediately below the bony crest.
3. Maximum compressive stresses were seen on the lingual aspect of the cortical bone, and these values were very close to approximating the ultimate strength of the bone.
4. Resin materials are beneficial in reducing the stresses endured under different loading conditions. Acrylic resin or glass-modified composite resin reduced the stress by 25% and 15%, respectively, when compared to equivalent thicknesses of porcelain or metal.

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